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Shoulder load during wheelchair-related activities of daily life

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ABSTRACT

Manual wheelchair users experience significant upper extremity strain, leading to a high prevalence of shoulder pain. Identifying modifiable risk factors for shoulder complaints is crucial for developing effective interventions. Consequently, it's important to quantify shoulder load (magnitude, frequency and duration) experienced by manual wheelchair users throughout the day.

This study aims to quantify the magnitude of shoulder load during various daily activities, including wheelchair propulsion at different speeds and inclines, ascending and descending ramps, weight relief lift, material handling and desk work. Ten able-bodied participants performed these activities while their upper extremity kinematics and exerted forces were measured. The analysis focused on glenohumeral contact force and rotator cuff muscle forces using the Delft Shoulder and Elbow Model.

Highest mean glenohumeral contact forces were found during weight relief lift (1363 ± 1204 N), followed by descending a ramp (997 ± 1043 N) and fast propulsion (802 ± 742 N). The supraspinatus muscle generated the greatest force during weight relief lift (327 ± 490 N) and fast propulsion (184 ± 205 N). These findings provide a first reference for estimating joint load in daily activities. By combining these data with the individual activity frequency and duration, personalized shoulder load exposure can be assessed, informing the development of targeted interventions to reduce shoulder pain in manual wheelchair users.

1. Introduction

Manual wheelchair users are dependent on their upper extremity for mobility and many other activities of daily living. These activities place a substantial load on the upper extremities (Morrow et al., 2010a, Rouvier et al., 2022), which results in a high prevalence of shoulder pain (39–44 % of wheelchair users (Bossuyt et al., 2024, Liampas et al., 2021)) and shoulder pathologies (49 % rotator cuff tears (Akbar et al., 2011), 42 %–80 % acromioclavicular joint arthrosis (Akbar et al., 2010, Arnet et al., 2021)). Since shoulder pain and pathology have a huge impact on physical activity and quality of life of the affected persons (Gutierrez et al., 2007), targeted interventions on risk factors for shoulder complaints are essential for the population of manual wheelchair users.

The identification of modifiable risk factors for shoulder complaints is essential to develop an intervention program. Risk factors can be divided into three domains: individual factors (posture, physical capacity, skills), environmental factors (wheelchair, ground type, inclination) and work requirements (magnitude, frequency and duration of load, recovery time) (Hastings and Goldstein, 2004). It has been recommended that the work requirements of manual wheelchair users related to shoulder complaints should be lowered as far as possible by minimizing the frequency and duration of repetitive upper limb tasks and force required to complete these tasks (Paralyzed Veterans of America Consortium for Spinal Cord Medicine, 2005). To give targeted recommendation, the manual wheelchair user's exposure to shoulder load (magnitude, frequency and duration) throughout the day has to be quantified.

Attempts have been made to quantify shoulder load exposure in manual wheelchair users in daily life conditions. Amrein et al. trained a neural network to predict shoulder load based on data from inertial measurement units (IMU) and electromyography (EMG) (Amrein et al., 2023). The comparison of this predicted shoulder load with the shoulder load determined by musculoskeletal modeling showed high similarity. This study demonstrated the feasibility of using wearable sensors and neural networks to estimate the shoulder load in wheelchair-related activities of daily living. However, even the sparse sensor setup (one IMU attached to the participant's upper arm, two IMUs attached to the

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wheelchair and two EMG sensors) is still complex and impractical and limits its usability for long-term real-life measurements.

Another approach to quantify shoulder load exposure in daily life is to monitor the performance of specific activities of daily living (information on frequency und duration), and assign the according shoulder load to each activity (magnitude). Several attempts have been made to monitor specific activities of daily living (ADL) by means of IMUs solely with the aim to evaluate **frequency and duration** of these activities. There are validated methodologies that quantify manual wheelchair propulsion in daily life (de Vries et al., 2023) or during sporting activities and match settings (van der Slikke et al., 2015). With two to three IMUs on the wheelchair, clinically relevant wheelchair mobility metrics, such as number and duration of pushes, number and magnitude of turns, or wheelchair velocity and inclination can be derived.

In the life of a wheelchair user, there are of course additional activities beyond wheelchair propulsion that contribute to shoulder loading. De Vries et al. developed and validated a method to identify the performance of wheelchair related ADL from wearable sensor data (de Vries et al., 2022). A generalizable algorithm could be trained by a deep learning model to reliably classify wheelchair related ADL from one IMU worn at the wrist. The classified activities covered a broad range of wheelchair related activities from daily life and included wheelchair propulsion (continuous or in restricted space), transfer, weight relief lift, manual material handling, deskwork, arm cranking and sitting still. The classification of a subset of these activities was also studied by Fortune et al. (wheelchair propulsion, non-propulsion activity, and static) (Fortune et al., 2022) or Skovbjerg et al. (wheelchair ambulation, sitting, lying, driving and other walking related activities) (Skovbjerg et al., 2022). By detecting a broad range of shoulder loading activities in the life of manual wheelchair users, the **frequency and duration** of these activities can be quantified.

The **magnitude** of shoulder load during wheelchair related tasks can be quantified by analyzing shoulder joint forces and moments or relative muscle activity of shoulder muscles (Rouvier et al., 2022). The total glenohumeral contact force gives a comprehensive indication of shoulder load, since it includes external forces and exerted muscles forces acting on the joint. Previous research has shown that upward directed forces at the shoulder are associated with increased signs of shoulder pathology (Mercer et al., 2006).

The magnitude of shoulder load expressed as glenohumeral contact force during level wheelchair propulsion has been extensively studied (Table 1). Fewer studies have reported glenohumeral contact force during wheelchair propulsion under different conditions, such as propelling up an incline, on a cross slope, during start and stop of wheelchair propulsion or during sprinting and dribbling in wheelchair basketball (Table 1). Also, alternative modes of propulsion have been investigated, such as handcycling or power-assisted wheelchair propulsion (Table 1). Other ADL often performed by wheelchair users are scarcely studied, with weight relief lift for pressure injury prevention and reaching being the only studied activities. There might be other ADL that are equally or even more challenging to the shoulder of wheelchair users. In the general or able-bodied population, a set of additional ADLs has been analyzed, such as glenohumeral contact force during reaching, lifting objects, walking with crutches, standing up from a chair, drinking, eating or combing hair (Table 1). Some of these activities are also relevant for wheelchair users, but not all of them are challenging for the shoulder.

With this study we aim to investigate shoulder load over a range of shoulder loading activities (SL-ADL) in the daily life of manual wheelchair users. The investigated activities include manual wheelchair propulsion with different velocities and inclines, ascending and descending a short ramp, wheelchair propulsion in restricted space, weight relief lift, manual material handling and deskwork. We hypothesize that the highest shoulder joint load is experienced during weight relief lift and the lowest during desk work.

In a future step, the above-mentioned SL-ADL can be monitored over a longer period in daily life to quantify frequency and duration. While assigning the magnitude of shoulder load to these activities, a shoulder load profile (exposure) can be established. This will help in identifying exposures that can cause shoulder injury and forms the basis for designing targeted interventions to lower shoulder pain.

Table 1
Previous studies reporting on the magnitude of shoulder load expressed as glenohumeral contact force during ADL specific for wheelchair users and general ADL.

Task	Studies
Manual wheelchair propulsion	
On level surface	Arnet et al., 2021; Collinger et al., 2008; Holloway et al., 2015; Kulig et al., 1998; Mercer et al., 2006; Morrow et al., 2010a; van Drongelen et al., 2005; Veeger et al., 2002
On inclined surface	Holloway et al., 2015; Kulig et al., 1998; Morrow et al., 2010a
On cross slope	Holloway et al., 2015
Starting and stopping	Morrow et al., 2010a
Sprinting and dribbling	Chenier et al., 2022
Alternative modes of propulsion	
Handcycling	Arnet et al., 2012
Power-assisted wheelchair	Kloosterman et al., 2012, 2015
ADL specific for wheelchair users	
Weight relief lift	Morrow et al., 2010a,b; van Drongelen et al., 2005
ADL not specific for wheelchair users	
Reaching	Bergmann et al., 2007; Charlton and Johnson, 2006; Klemt et al., 2018; van Drongelen et al., 2005
Lifting objects	Anglin et al., 2000; Bergmann et al., 2007; Charlton and Johnson, 2006; Klemt et al., 2018
Walking with crutches	Anglin et al., 2000; Bergmann et al., 2007; Klemt et al., 2018
Standing up from chair	Anglin et al., 2000; Klemt et al., 2018
Eating	Charlton and Johnson, 2006; Klemt et al., 2018
Combing hair	Bergmann et al., 2007; Charlton and Johnson, 2006; Klemt et al., 2018

2. Methods

2.1. Participants

Ten able bodied participants were included in the study (7 female; age 39 ± 9 years; height 1.69 ± 0.09 m; weight 66 ± 12 kg). Each participant completed a training session in the wheelchair one week prior to the experiments. They trained all SL-ADL of interest (Table 2) until they were comfortable with the performance and executed the activities in a smooth manner.

Ethical approval for this study was obtained from the Ethikkomission Nordwest- und Zentralschweiz (EKNZ), project-ID 2022–01961 and all participants provided written informed consent prior to participation.

2.2. Experimental trials

Participants were invited to the Movement Analysis Laboratory of Swiss Paraplegic Research. After preparing the participants and completing the calibration measurement, they were asked to perform a given range of SL-ADL (Table 2). The selection of SL-ADL was based on literature results and expert opinion on clinically relevant shoulder loading tasks.

The participants were instructed on the activities to perform, but they were free to execute the activities in their preferred manner and velocity, except for the given velocity on the treadmill.

Table 2
List of shoulder loading activities of daily life (SL-ADL) performed by the participants.

SL-ADL	Description
Slow wheelchair propulsion (WCprop_slow)	Continuous wheelchair propulsion (30 s) on the treadmill at 0.56 m/s at 0 % inclination.
Fast wheelchair propulsion (WCprop_fast)	Continuous wheelchair propulsion (30 s) on the treadmill at 1.11 m/s at 0 % inclination.
Slow wheelchair propulsion at incline (WCprop_incline)	Continuous wheelchair propulsion (30 s) on the treadmill at 0.56 m/s at 6 % (3.4°) inclination.
Maneuvering	Intermitted wheelchair propulsion in restricted space: maximum three meters distance covered, maximum 3 consecutive pushes, including turns and backwards propulsion.
Ascending a ramp (Ramp_asc)	Driving with the wheelchair up a short ramp (12 % (6.8°) incline, length 0.4 m), starting from standstill right in front of the ramp.
Descending a ramp (Ramp_desc)	Driving with the wheelchair down a short ramp (12 % (6.8°) incline, length 0.4 m), starting from standstill right in front of the descent.
Weight relief lift (WRL)	Weight relief lift for pressure injury prevention: starting with placing the hands on the rim of the wheelchair wheel, then push up and lift the bottom off from the seat cushion, hold for ten seconds and release to sit.
Manual material handling (MMH_front)	Collect and place a weight of 2 kg to cupboard shelves of four different heights (0.35 m, 0.8 m, 1.1 m and 1.4 m above ground): collect the weight from the lowest shelf and place it to the next higher shelf. Collect the weight again and place it on the next higher shelf. When reaching the highest shelf continue with placing the weight to the next lower shelf until reaching the lowest shelf.
Manual material handling at the back (MMH_back)	Placing a weight of 2 kg into the back pocket of the wheelchair: starts with holding the weight in the hand, then placing it into the back pocket.
Deskwork	Sitting at a desk and performing deskwork (30 s): typing on a keyboard, using the computer mouse and the mobile phone.

2.3. Data collection and processing

All measurements were performed in a standard active wheelchair (Küschall Compact 2017, Küschall AG, Witterswil, Switzerland). For collecting propulsion kinetics, the wheelchair was fitted with a SmartWheel (24 in.; Three Rivers Holdings LLC, Mesa, Arizona, USA) on the right side and a dummy wheel on the contralateral side. The SmartWheel recorded the external forces at 240 Hz. Kinetic data were offset corrected, filtered with a fourth order Butterworth filter with a cutoff frequency of 20 Hz and resampled to 100 Hz for further input to a musculoskeletal model.

Kinematics of the upper extremity were captured with an eight-camera movement analysis system (Oqus, Qualisys AB, Gothenburg, Sweden) operating at 100 Hz. Unique clusters of reflective markers were placed on the trunk and the right upper extremity (thorax, acromion, upper arm, forearm and hand). Prior to the experiment, calibration measurements with a pointer were performed to define the relevant bony landmarks (van der Helm, 1997) relative to the cluster markers. With this relationship, the positions of the anatomical landmarks during the actual experiment were reconstructed from the recorded cluster markers, as well as the local coordinate systems of the thorax, clavicle, scapula, humerus and forearm (Wu et al., 2005). The position of the glenohumeral joint was calculated based on the regression equation based on Meskers (Meskers et al., 1998). The point of force application during wheelchair propulsion was defined as the midpoint between radial and ulnar styloid. Kinematic data were filtered with a fourth order Butterworth filter with a cutoff frequency of 6 Hz.

2.4. Musculoskeletal model

To analyze shoulder load, the individual upper body kinematics and external forces were input to the Delft Shoulder and Elbow Model (DSEM)(Nikooyan et al., 2011). The DSEM is a large-scale inverse-dynamics based model in which anatomical structures are modelled by mechanical elements. It includes all bones, joints, and most ligaments of the shoulder as well as 31 muscles divided into 139 muscle elements, resulting in 17-DOF. This comprehensive geometry is based on cadaveric measurements (Klein Breteler et al., 1999). Input to the model were thorax orientation and joint angles of the scapula, clavicula, upper- und forearm, as well as external forces (SmartWheel data for WCprop, maneuvering, ascending and descending the ramp and WRL, and a known weight of 2 kg for manual material handling). From this input, the clavicular and scapular orientation are optimized so the scapular medial border stays attached to the thorax and the conoid ligament stays at equal length (Nikooyan et al., 2011, van der Helm, 1994a, van der Helm, 1994b). The individual muscle forces are calculated based on an energy cost function (Praagman et al., 2006). Muscle forces were calculated as the sum of the forces applied by each muscle element. Glenohumeral contact force (GHCF) was calculated by the vector summation of the model-estimated muscle forces around the glenohumeral joint and the external force. A joint stability constraint is included in the model to guarantee that the resulting joint contact force is directed into the glenoid of the scapula.

2.5. Data analysis

Shoulder load of the different SL-ADL was analyzed descriptively by mean and peak values of GHCF and muscle forces exerted by the rotator cuff muscles. Mean values were calculated over either the whole tasks (wheelchair propulsion, maneuvering, ascending/descending a ramp, manual material handling, deskwork) or over two repetitions (weight relief lift). Peak values are calculated accordingly, taking the average over the highest 10 % values of each task.

In addition to mean and peak forces, the force distribution during each activity was analyzed. For this the occurring GHCF and muscle forces per frame were assigned to bins of 100 N (for GHCF) and 10 N (for muscle forces). The proportion of forces occurring in each bin is depicted in a proportion histogram.

3. Results

3.1. Glenohumeral contact force

Shoulder load varied a lot over the different SL-ADL. A typical example of the distribution of the GHCF is displayed in Fig. 1.

The mean ± SD GHCF of WRL was the highest for all activities (1363 ± 1204 N), followed by descending the short ramp (997 ± 1043 N) and fast wheelchair propulsion (802 ± 742 N). The peak forces were generally two to three times higher than the mean values. Also, peak forces were highest for WRL (3984 ± 612 N), followed by descending (3487 ± 919 N) and ascending the short ramp (2627 ± 540 N). The mean and peak GHCF values of all analyzed SL-ADL are given in Table 3.

Fig. 2 shows that, during most activities, over 50 % of the glenohumeral contact forces fall within the range of 0 to 500 N, with a rapid decrease beyond this range. The biggest exception is seen in WRL, where the force distribution is more even.

3.2. Muscle force of the rotator cuff

The model-estimated supraspinatus muscle produces the highest mean force during WRL (327 ± 490 N) and fast wheelchair propulsion (184 ± 205 N). Highest mean forces of the infraspinatus were observed during fast wheelchair propulsion (396 ± 383 N) and WRL (313 ± 425 N). Subscapularis produces the highest mean force during ascending

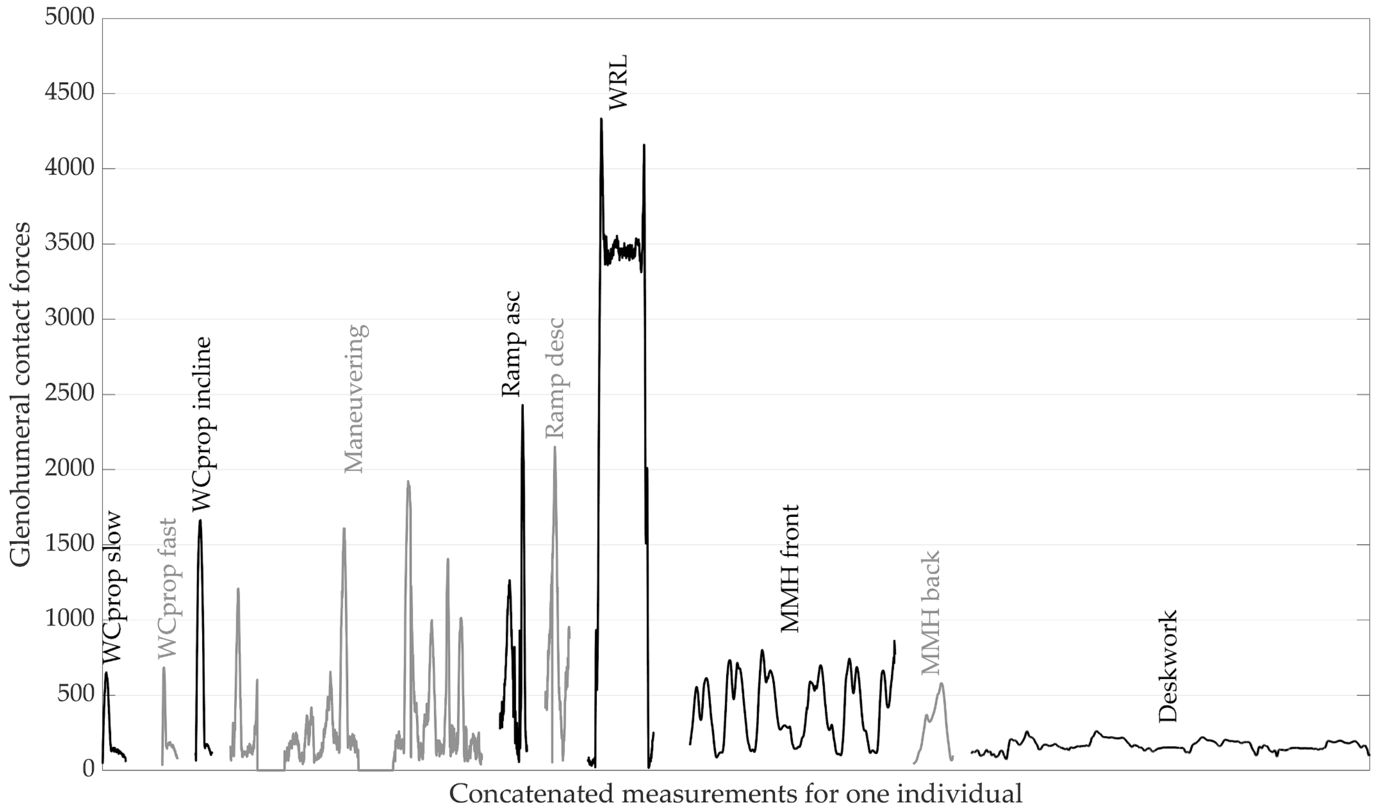

Fig. 1. Typical example of glenohumeral joint reaction forces during all measured activities. WCprop_slow = slow manual wheelchair propulsion (0.56 m/s, 0 % inclination), WCprop_fast = fast manual wheelchair propulsion (1.11 m/s, 0 % inclination), WCprop_incline = slow manual wheelchair propulsion on an incline (0.56 m/s, 6 % inclination), Maneuvering = intermitted wheelchair propulsion in restricted space, Ramp_asc = ascending ramp, Ramp_desc = descending ramp, WRL = weight relief lift, MMH_front = manual material handling in front, MMH_back = manual material handling in back.

(268 ± 458 N) and descending the ramp (242 ± 459 N). Teres minor contributed with less force, reaching highest force during descending (14 ± 29 N) and ascending the ramp (13 ± 25 N). The mean and peak muscle forces of all analyzed SL-ADL are given in Table 3.

Fig. 3 shows the contribution of the different rotator cuff muscle to the analyzed activities. During activities like deskwork or slow wheelchair propulsion it's mainly the infraspinatus that produces the highest muscle forces. For other activities, like WRL or maneuvering, the muscle contribution is more evenly distributed over the rotator cuff muscles.

4. Discussion

The most demanding activity for the shoulder complex is **WRL**. During this activity, the highest GHCF, representing the total load on the shoulder, as well as the highest supraspinatus and infraspinatus forces, were observed (Table 3). These high values are not surprising since the whole upper body must be pushed off the wheelchair. The GHCF values are higher compared to previous reported mean values of 648 N for a WRL (van Drongelen et al., 2005, van Drongelen et al., 2011). The same accounts for muscle forces of the rotator cuff (van Drongelen et al., 2005). The differences might be attributed to adaptions made to the DSEM within the last twenty years, like different cost functions for calculating muscle contribution, such as the energy cost function used in the present study. Whether the high forces also result in damage cannot be conclusively determined. From previous research it is known that performing a WRL is reducing the subacromial space, which is seen as one of the risk factors for subacromial pain or damage (Arnet et al., 2022). Next to the individual shoulder capacity to stabilize the joint during this weight bearing task, the frequency of WRLs per day is determining the risk for shoulder problems. So far, only limited data are available on WRL performance over the day. Sonenblum et al. reported that wheelchair users performed on average four WRL per day (Sonenblum et al., 2016). However, more observational studies are needed to estimate the daily shoulder load of WRL in wheelchair users.

Overcoming short but steep **ramps**, as it can happen mainly in outdoor ambulation, is a further activity which is demanding to the shoulder joint (Table 3). Both, ascending and descending the ramp is an activity in which high forces must be applied by the hand to the rim in a short time, either for pushing up the inclination or for breaking. For ascending the ramp, mainly the subscapularis is active to internally rotate the humerus. While descending and breaking, teres minor is contributing to the retroversion of the humerus. Overcoming short and steep ramps is needed for example for entering or exiting public transport or buildings. Depending on the lifestyle or the job of wheelchair users, the number of ramps to overcome during the can vary a lot. So far, no data are available on the number of ramps to overcome in real life and the corresponding cumulative load on the shoulder.

The shoulder load during manual wheelchair propulsion is dependent on the inclination of the surface and the continuity of the propulsion, but not on speed (Table 3). **Slow wheelchair propulsion** results in a mean GCHF of 302 N per push. This is the second-lowest GHCF among all the activities analyzed. Previous studies have reported slightly lower mean and peak values of 182 N and 295 N respectively (van Drongelen et al., 2011). A recent study performed in the daily life of 19 manual wheelchair users stated that there is a high variance in the number of pushes performed per day. The number of pushes varied from 438 to 4820, with an average over participants of 2055 pushes (de Vries et al., 2024). Although shoulder load per push is low, the cumulative load from daily pushes can increase the risk of shoulder overload.

Fast wheelchair propulsion does not result in a relevant increase of shoulder load compared to level propulsion with the same speed. GHCF magnitude as well as muscle forces are comparable for both conditions

Table 3
Shoulder load of the different activities, quantified by the GHCF and the muscle forces exerted by the rotator cuff muscles. Values are given as mean and peak values over the activity and the standard deviations in brackets. Colors indicating the order of magnitude of the shoulder load values separately for each component (mean GHCF, max GHCF etc.). The darker the red the higher are the values.

	GHCF [N]		Supraspinatus [N]		Infraspinatus [N]		Subscapularis [N]		Teres minor [N]	
	Mean	Max	Mean	Max	Mean	Max	Mean	Max	Mean	Max
WCprop_slow	302 (212)	751 (126)	59 (57)	197 (42)	127 (113)	369 (68)	11 (15)	47 (10)	3 (6)	18 (6)
WCprop_fast	312 (277)	980 (153)	58 (71)	235 (73)	120 (141)	481 (82)	15 (22)	69 (16)	4 (9)	26 (13)
WCprop_incline	802 (742)	2293 (371)	184 (205)	617 (128)	369 (383)	1130 (87)	44 (61)	181 (44)	8 (17)	50 (25)
Maneuvering	464 (445)	1505 (565)	63 (68)	225 (79)	117 (168)	520 (203)	105 (216)	612 (378)	8 (21)	60 (31)
Ramp_asc	791 (778)	2627 (540)	90 (94)	314 (93)	200 (315)	991 (258)	268 (458)	1436 (399)	13 (25)	77 (27)
Ramp_desc	997 (1043)	3487 (919)	140 (192)	597 (291)	245 (430)	1184 (793)	241 (459)	1442 (555)	14 (29)	92 (26)
WRL	1363 (1204)	3984 (612)	327 (490)	1646 (366)	313 (425)	1346 (333)	203 (329)	1031 (467)	11 (31)	95 (42)
MMH_front	399 (213)	766 (46)	59 (34)	124 (13)	115 (74)	261 (27)	28 (26)	85 (23)	12 (18)	56 (11)
MMH_back	321 (216)	752 (117)	40 (32)	105 (17)	94 (71)	221 (26)	17 (41)	101 (89)	12 (20)	60 (13)
Deskwork	249 (52)	337 (12)	35 (8)	48 (2)	76 (17)	100 (3)	24 (15)	53 (6)	4 (4)	14 (2)

WCprop_slow = slow manual wheelchair propulsion (0.56 m/s, 0 % inclination), WCprop_fast = fast manual wheelchair propulsion (1.11 m/s, 0 % inclination), WCprop_incline = slow manual wheelchair propulsion on an incline (0.56 m/s, 6 % inclination), Maneuvering = intermitted wheelchair propulsion in restricted space, Ramp_asc = ascending ramp, Ramp_desc = descending ramp, WRL = weight relief lift, MMH_front = manual material handling in front, MMH_back = manual material handling in back.

(Table 3). The small difference in shoulder load between slow and fast propulsion might be a result of change in propulsion technique. If individuals increase push frequency at fast speed, as done by the participants of this study (slow: 49 pushes/min, fast: 68 pushes/min), shoulder load per push is reduced compared to individuals who keep push frequency steady. However, previous studies have reported that joint forces more than double when speed increases from 1.5 m/s to 2.3 m/s (Kulig et al., 1998). Such high velocities are rarely seen in daily life. Wheelchair propulsion speed does vary between persons and daily activities, but weekly averages are much lower, ranging 0.43 to 0.88 m/s (Wilson et al., 2008). Our data suggest that propulsion speed is not a critical factor in estimating shoulder load in daily life based on performed activities.

Not surprisingly, propelling up an **incline** of 6 % at slow speed is much more demanding to the shoulder joint than level propulsion at the same speed. GHCF and rotator cuff muscle forces are approximately three times higher while propelling up the incline (Table 3). Similar increases in joint forces have been found in previous studies; three times higher peak GHCF when propelling on an incline of 6.5 % (Holloway et al., 2015) and 3.5 times higher mean and peak GHCF on an incline of 8.3 % incline (Morrow et al., 2010b) and 2.2 higher peak GHCF on an incline of 8 % (Kulig et al., 1998). All muscles are producing higher forces, and mainly the infraspinatus is adding to the high joint load (Table 3). If one rotator cuff muscle is producing high forces over a longer time, this could lead to local fatigue, possibly resulting in a less stable shoulder joint. Therefore, the amount of incline propulsion should be quantified when estimating shoulder load over a day. However, a recent study has shown that manual wheelchair users rarely propel longer distances over inclinations of more than 3° and inclines higher than 6° are seldom observed (de Vries et al., 2024).

Adding starts, stops and turns to slow propulsion, as happening during the activity called **maneuvering**, increases shoulder load (Table 3). Mainly subscapularis is producing much more force than during slow propulsion. Maneuvering includes intermitted wheelchair propulsion in restricted space and represents mostly indoor activities like working in an office, preparing a meal in the kitchen or doing household activities. So far, no information is present in literature on number of start and stops performed in daily life of wheelchair users. Turns (larger than 30°) are performed frequently. On average, 635 turns were detected over a day, with a minimum of 269 and a maximum of 1396 turns (de Vries et al., 2024). In the present analysis we did not distinguish between shoulder load of a turn, start and stop, since turning is often performed in combination with starting and stopping the wheelchair, but analyzed the activity "maneuvering" as a whole. The high number of turns performed in daily life suggests that the cumulative load of maneuvering will be high when analyzed over the day.

Manual material handling in front (collect and place a weight of 2 kg to cupboard shelves) and in the back (placing a weight of 2 kg into the back pocket) is not so demanding for the shoulder joint (Table 3). Comparable peak GHCF measured with an instrumented prosthesis have been found while putting 2.5 kg into a shelf (677 N, (Bergmann et al., 2007)). Information on the frequency of manual material handling in daily life is as yet unavailable.

Performing **deskwork** is the activity that is least demanding for the shoulder joint. Mean GHCF of 249 N are present during this activity and rotator cuff muscles are not highly active (Table 3). Since persons with SCI often change to office jobs after becoming wheelchair dependent (Schwegler et al., 2021), deskwork activity might be present in most working persons with SCI. Even a low joint load can accumulate over time, depending on the workload, further contributing to the already increased load from wheelchair-related daily activities.

The presented data on joint load magnitude provides a basis for

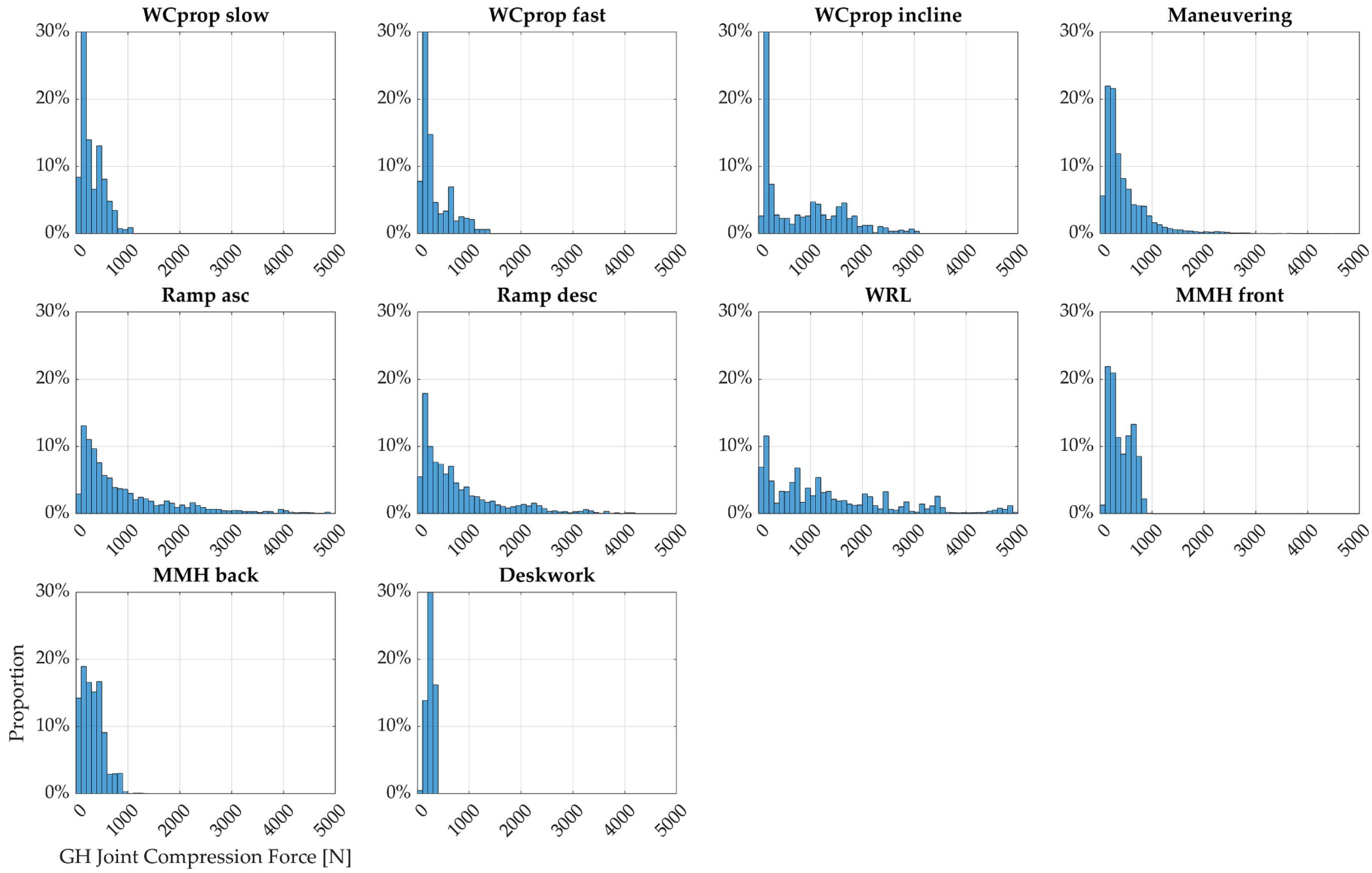

Fig. 2. Distribution of glenohumeral contact force acting during the different activities.

estimating joint load in daily life. By monitoring the studied activities using IMUs over an extended period, their frequency and duration can be quantified, allowing the assigned joint load magnitudes to be accumulated. This allows for quantification of exposure to shoulder load due to daily activity. Monitoring concurrently shoulder load exposure and progress of shoulder problems, such as injury or pain, enables examination of causality and forms the basis for identifying targets of intervention to lower the high prevalence of shoulder pain in manual wheelchair users.

5. Limitations

The analyzed activities were selected based on literature and expert opinion and cover therefore a broad spectrum of shoulder loading activities of wheelchair dependent persons. We are however aware of the fact that this list is not complete, and some relevant activities might be missing. One prominent activity that is missing on this list is transferring. Especially a transfer to a non-level surface or to a more distant place, as transferring to a car seat, will be very demanding to the shoulder joint. Due to technical difficulties, like quantifying the external load of during a transfer, this activity is not included in the study. It might also be that we missed other relevant activities, since they are not identified by the experts so far. This potential gap in knowledge could for example be closed by measuring with an instrumented shoulder prosthesis over a longer period in the daily life of a manual wheelchair user.

Another limitation of the study is the inclusion of able-bodied participants. The data used for this study was collected within a bigger project with the aim to develop a methodology for directly assessing shoulder load based on IMUs and EMG (Amrein et al., 2023). For method development we opted or able-bodied participants since recruitment success in this population is more promising and the choice of the population did not bias method development. A previous study has shown that mean and peak GHCF during wheelchair propulsion and WRL do not differ between able-bodied participant and participants with paraplegia. However, joint forces of persons with tetraplegia are higher (van Drongelen et al., 2011). To avoid bias due to lesion level, it would be ideal to repeat this study with persons with a variety of lesion levels.

A third limitation relates to the standardization of the performed activities, which is not concurrent with daily life situations. For most of the activities the participants were clearly instructed how to perform the task. Also, wheelchair propulsion on the treadmill was standardized on speed and only straight propulsion was possible. Only during the deskwork-task the participants were free to choose the duration of the different tasks (typing, using computer mouse or phone). In real life there is a high variability in these tasks which is not included in the highly standardized performance during the laboratory measurements. It would be ideal to add more variation also for the lab measurements to better represent daily life activities.

6. Conclusions

This study examined joint load, defined as GHCF and rotator cuff muscle forces, across various shoulder-loading activities relevant to individuals with SCI. Performing WRL and overcome level differences, either by propelling on inclined surface or ascending and descending short ramps, place the highest load on the shoulder. These activities are important for living an independent life as a manual wheelchair user, either for selfcare (WRL) or for outdoor mobility.

The presented data provide a first reference for estimating joint load in daily life. By combining these reference data on magnitude of joint

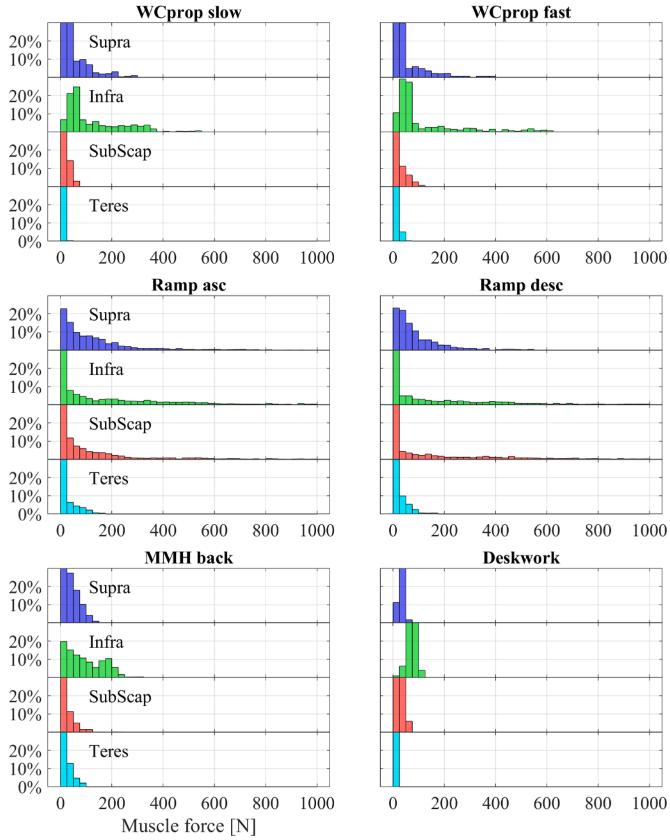

Fig. 3. Distribution of rotator cuff muscle forces acting during the different activities. Supra = supraspinatus, Infra = infraspinatus, SubScap = subscapularis, Teres = teres minor.

load for specific activities with individually measured frequency and duration of these activities, the personal shoulder load exposure can be determined. In combination with information on progression of shoulder problems, causality can be inferred, which forms the basis for developing targeted interventions to reduce shoulder pain in manual wheelchair users.

CRediT authorship contribution statement

Ursina Arnet: Conceptualization, Formal analysis, Methodology, Resources, Supervision, Validation, Writing – original draft, Writing – review & editing. **Dirkjan (H. E. J.) Veeger:** Formal analysis, Software, Writing – review & editing, Methodology. **Wiebe H. K. de Vries:** Conceptualization, Investigation, Methodology, Project administration, Writing – review & editing.

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Declaration of competing interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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